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SUBSTITUTE SPECIFICATION

MAGNETIC RESONANCE IMAGING DIAGNOSTIC APPARATUS AND METHOD THEREFOR

FIELD OF THE INVENTION

The present invention relates to a magnetic resonance imaging diagnostic apparatus (hereinafter simply referred to as an MRI apparatus), and, more particularly, to an MRI apparatus having a spectroscopic imaging function.

BACKGROUND OF THE INVENTION

A spectroscopic imaging method using a magnetic resonance diagnostic apparatus for medical use (MRI) (hereinafter referred simply as an to MRSI method) is a method used for imaging a distribution of specific chemical components in the living body. This method can acquire not only anatomical information, which is acquired in an ordinary MRI, but it also can acquire chemical information concerning carbohydrate metabolism or energy metabolism or the like. So, it is useful for diagnosis at the first stage of a disease.

Various kinds of MRSI methods have been attempted, such as a 3D-CSI method for measuring a chemical shift as a temporal coefficient, but a method using a proton (^1H nucleus) is generally used in an MRI apparatus for clinical use. The 3D-CSI method includes phase encoding of spatial information in the directions x and y, and spectrum information is encoded with the progress of time in a signal measurement.

In an MRSI method to measure a proton, signals due to water

and fat disturb the measurement because they are included in the living body by more than 2 to 4 orders in comparison with metabolic materials. So, it is necessary to additionally employ a technique to suppress this disturbance. Prior art will be described with reference to Fig. 9 and Fig. 10. In the prior art, the spectrum of water is selectively excited in preprocessing by a CHESS pulse 31, and, subsequently, a gradient magnetic field 33 is applied to dephase the magnetization of water, so as not to produce a signal. Furthermore, after having excited the slice perpendicular to the imaging section (fat part) with a pulse 32, a gradient magnetic field 34 is applied to dephase this magnetization in this slice (OVS : Outer Volume Suppression). This process is repeated to cover all of the fat region while changing the slice. After this preprocessing is completed, a spectroscopic imaging is performed.

An example will be described using a transaxial slice of a person's head with reference to Fig. 9. Excitation and dephasing are performed from slice 1 to slice 8 while changing the direction so as to cover the head surface in an area with nearly an ellipse shape. Eight slices are used in order to cover the subcutaneous fat in Fig. 9; however, usually 4 to 8 slices are used.

In this traditional fat suppression method, it was difficult to cover a fat region precisely because a fat system is covered with a plural number of rectangles. And, it is difficult with this method to fully suppress the signal

produced from fat. On the other hand, if it becomes necessary to increase the number of rectangles for fully covering the fat system, it is troublesome to set the rectangular region. In addition, if the number of rectangles is increased, the preprocessing will take a longer time. And, recovery of water magnetization will be generated due to longitudinal relaxation, and the suppression of the influence of water will be insufficient.

Furthermore in the 3D-CSI method, at least a double phase encode loop is used because spatial coordinate information of 2 or 3 axes are given in the signal, so it takes a long time to complete the measurement. But, in a traditional fat suppression method, it takes about 20 to 50 ms for preprocessing. So, this is also a problem in that the imaging time is elongated.

Thus, the object of the present invention is to provide an MRI apparatus that is capable of realizing an MRSI method in which a useless signal is sure to be suppressed, and there is no elongation of the measurement time.

SUMMARY OF THE INVENTION

In order to solve such problems as described above, in accordance with this invention, an MRI apparatus for performing spectroscopic measurement includes a function which involves the application of a swing gradient magnetic field in the direction of two or three axes, together with a predetermined high frequency magnetic field (RF). This makes no use of a fat suppression process in the phase encode loop,

and a signal from the fat system can be greatly reduced.

A technique for performing a spatial selective excitation with the combination of a swing gradient magnetic field and RF wave form calculated from a form function of excitation are precisely described in an article entitled "A k-Space Analysis of Small-Tip-Angle Excitation", in J.Magn.Reson., 81, 43-56 (1989) or the like, by the author J. Pauly, D. Nishimura and A.Macovski. But, an MRI apparatus of the present invention is designed to provide means for applying techniques of this spatial selective excitation concretely to an MRSI measurement.

That is to say, an MRI apparatus of the present invention comprises magnetic field generation means for generating a static magnetic field, a gradient magnetic field and a high frequency magnetic field (RF) to be applied to the object to be examined, respectively; detection means for detecting a magnetic resonance signal generated from the object to be examined; image reconstruction means for reconstructing an image with a detected magnetic resonance signal; display means for displaying the reconstructed image; and control means for controlling each means.

The control means comprises means for setting a desired region of interest (ROI) in a predetermined region of the object, and means for calculating the modulation wave form of RF excited selectively in this ROI. And, the control means controls the application of an RF pulse modulated with the modulation wave form and a swing gradient magnetic field in

the direction of two or three axes simultaneously, and acquires signals including spectrum information about the predetermined region of the object.

Only the interior of the ROI with an arbitrary shape that is being looked at is excited and the signal is acquired therefrom. So, the effect of fat and water from the exterior of the ROI can be effectively suppressed. Thus, a fixed quantity of the spectrum can be improved and the accuracy of diagnosis can be improved. In addition, the excitation pulse is selective in the region, so that there is no need to irradiate a fat suppression RF pulse in preprocessing for performing phase encoding repeatedly. Thus, the time for preprocessing can be shortened. As a result, recovery of a water signal can be reduced.

In the MRI apparatus of the present invention, the means for setting the ROI displays a slice image obtained at a desired slice position as a scout image, and operates interactively to set the ROI on the displayed scout image.

The means for calculating an RF waveform concretely determines an excitation form function from the shape of the ROI, and calculates a modulated waveform of a high frequency magnetic field by two-dimensional Fourier transformation of this excitation form function in nearly real time.

This RF waveform for selective excitation is different from the shape of the ROI. But, by previously imaging a scout image and displaying it, the ROI is able to be set interactively on this scout image. Accordingly, an arbitrary

ROI shape can be easily set. In addition, a modulated RF waveform is generated by performing two-dimensional Fourier transformation with a computer to set the shape of the ROI. So, an RF waveform corresponding to the ROI can be acquired in nearly real time, and, an MRSI measurement can be performed soon after the setting of the ROI.

Moreover, in a preferred embodiment of an MRI apparatus according to the present invention, control means sets the smallest rectangular region containing the ROI, and determines an image matrix size according to this rectangular region.

The image matrix size is the size of image data for expressing the number of pixels (or voxel), which is the number of lines × rows, arranged in a two-dimensional or three-dimensional array. It is defined by the number of gradient magnetic fields encoded for encoding spatial information. Accordingly, there is no need to make the size of image matrix so large, because it is defined by the smallest rectangle capable of covering the shape of the ROI. As a result, there is no need to repeat useless phase encoding, and the measurement time of the MRSI can be shortened.

BRIEF DESCRIPTION OF THE DRAWINGS

Fig. 1 is a flow chart showing a method according to one embodiment of the present invention.

Figs. 2(a) to 2(d) are diagrams which show a method of setting an ROI interactively, and an matrix.

Fig. 3 is a timing diagram showing one example of a sequence applied to the present invention.

Fig. 4 is a schematic block diagram showing an example of the MRI apparatus of the present invention.

Fig. 5 is a timing diagram showing another example of a sequence applied to the present invention.

Fig. 6 is a flow chart showing the procedure according to another embodiment of the present invention.

Fig. 7 is a timing diagram showing another example of a sequence applied to the present invention.

Fig. 8 is a timing diagram showing another example of a sequence applied to the present invention.

Fig. 9 is a diagram showing a traditional fat suppression method.

Fig. 10 is a timing diagram showing a sequence with the combination of water suppression and fat suppression in traditional type method.

THE BEST MODE FOR CARRYING OUT THE INVENTION

The present invention will be described precisely with reference to Fig. 1 or Fig. 8.

Fig. 4 is a schematic block diagram of an MRI apparatus which employs the present invention. This MRI apparatus comprises a static magnetic field generating magnetic circuit 402 for generating a uniform static magnetic field B0 in the interior of the object 401 to be examined; a gradient magnetic field generation system 403 for generating gradient magnetic fields Gx, Gy, Gz and for controlling the intensity thereof linearly to three directions x, y and z that are perpendicular in each other; a transmitter system 404 for generating a high

frequency magnetic field; a detection system 405 for detecting a nuclear magnetic resonance signal generated from the object; a signal processing system 406; a computer 408 for performing calculation of image reconstruction and for controlling the whole apparatus; a sequencer 407 for controlling the gradient magnetic field generation system 403; a transmitter system 404; and a detection system 405 responsive to a command from computer 408 and operation part 421 for sending a necessary order to signal processing system 406 and computer 408.

The static magnetic field generation magnetic circuit 402 comprises an electromagnet or a permanent magnet. And, the object 401 is accommodated in the space of its static magnetic field.

The gradient magnetic field generation system 403 comprises a gradient magnetic field coil 409 with three axes, and a power supply 410 for supplying current to the gradient magnetic field coil. And, it generates a linear gradient magnetic field for applying spatial information to the magnetic resonance signal and a swing gradient magnetic field, when an ROI is selectively excited.

The transmitter system 404 comprises a synthesizer 414 for generating predetermined high frequency pulses, a modulator 412 for modulating a high frequency pulse generated from synthesizer 411, electric power amplifier 413, and a transmitter coil 414a. The modulation waveform modulated by this modulator 412 is calculated and saved by computer 408 as the following procedure and is supplied from sequencer 407. A

high frequency magnetic field is generated in the interior of the object 401 to excite a nuclear spin, by supplying a high frequency pulse modulated with a predetermined modulation waveform to transmitter coil 414a. An excited nuclear spin usually will use ^{1}H. But, other nuclear elements, such as ^{31}P, ^{12}C or the like also can be used.

The detection system 405 comprises a receiver coil 414b for receiving nuclear magnetic resonance signals emitted from the object 401, an amplifier 415, a quadrature phase detector 416, and an A/D converter 417. Nuclear magnetic resonance signals, which are received with receiver coil 414b and amplified with amplifier 415, are input to computer 408 after quadrature detection and A/D conversion.

The transmitter coil 414a and the receiver coil 414b can be separated as shown in the figure. But, a combined coil also can be used.

Computer 408 reconstructs an image corresponding to the density distribution of nuclear spin, the distribution of the relaxation time, the spectrum distribution or the like, and this information is displayed on the CRT display 428. Operation part 421 is composed within computer 408 and inputs a command, which is necessary for performing a program of processing in this MRI apparatus and for applying various kinds of settings to computer 408. Processing performed through this operation part 421 is carried out for setting the ROI (region of interest) to the predetermined region of the object, for determining an image matrix size in accordance

with the ROI, and for calculating the RF modulation waveform generated in the transmitter system 404 based on the shape of the ROI. Data in the middle or at the end of the calculation performed in this computer 408 is accommodated in memory 424 and memory 425.

Computer 408 controls the gradient magnetic field generation system 403, the transmitter system 404, and the detection system 405 through sequencer 407 to perform the imaging with a predetermined imaging sequence.

Next, an MRSI measurement performed with these MRI apparatus will be described. In the following description of the embodiment, it will be assumed that proton MRSI is performed on the transaxial face (x-y face) of an abdominal region. But, another slice can be applied to this method. In addition, suppression of fat will be described as an example, but it is also possible to suppress the spectrum other elements than fat.

Fig. 1 is a flow chart of the MRSI method of the present invention. Figs. 2(a) to 2(d) are diagrams showing the steps in setting of the ROI. Fig. 3 is a view showing one embodiment of imaging sequence used in the MRSI method of the present invention.

In the MRSI measurement performed in this embodiment, a scout image is acquired relative to the desired slice before a spectroscopic measurement is performed (Fig. 1, step 11). A scout image can be acquired by using a general imaging method of MRI, such as the FSE (Fast Spin Echo) method and the EPI

(Echo Planner Imaging) method. This selected slice is an arbitrary slice included within the region of an object to be measured.

After having displayed this scout image for display, an ROI in the form of a closed region is set with a mouse or the like so as to cover the desired region entirely without covering fat (step 12). Fig. 2(a) is a view showing typically the scout image displayed, and area 61 is an ROI input with the mouse.

Thus, after smoothing the set ROI 61, a binary value function, with the interior of the ROI being 1 and the exterior of the ROI being 0, is made to define form function D(x,y) (step 13). The shape of the ROI can be specified with an arbitrary shape in principle. However, as the shape is more complex, the Fourier transformation D'(kx,ky) of form function D(x,y) includes more high frequency components that represent the edge of a k space component. So, it is necessary to enlarge the k space in the imaging after that. This means that the amplitude of the swing gradient magnetic field is large, and its application time is long. In this case, it is difficult to provide equipment and the imaging time has to be expanded. Accordingly, in inputting the shape of the ROI, it is more practical to avoid making the ROI complex unnecessarily. In addition, as the borderline of the form function changes more sharply in rising up, fat region is suppressed sharply. But, it is severe in the amplitude characteristic and transient characteristic of the gradient

magnetic field. It is preferable to apply smoothing for the binary value function $D(x, y)$ necessarily. For example, on the borderline of the value D changing from 0 to 1, it has a certain amount of width to decrease the value gradually from 1 to 0.

Next, two dimensional Fourier transformation is performed on the form function $D(x,y)$ to generate the function $D'(kx,ky)$ of the k space. By using this function D', the RF waveform $B_1(t)$ is calculated with an equation (1) (step 14).

$$B_1(t)=D'(k_x(t),k_y(t))\cdot|\gamma G(t)| \quad \ldots(1)$$

The gradient magnetic field waveforms Gx(t)?Gy(t) are predetermined to an imaging sequence of the imaging. For example, the gradient magnetic field waveforms Gx(t), Gy(t) having the shape of a spiral orbit converging from the edge of the k space to the starting point with constant speed can be used. Specific examples of these gradient magnetic field waveforms Gx(t),Gy(t) are represented in equation (2). These are sine wave or cosine wave functions, in which their amplitudes decrease linearly as a function of time.

$$Gx(t) = -A/\gamma T[2\pi n(1 - t/T)\sin 2\pi nt/T + \cos 2\pi nt/T]$$

$$Gy(t) = A/\gamma T[2\pi n(1 - t/T)\cos 2\pi nt/T - \sin 2\pi nt/T] \quad \ldots(2)$$

In an equation (2), T/n is a rotation period of the spiral of said space. In addition, A is a fixed number for defining the size of the spiral. n is usually about 10.

In addition, the gradient magnetic field waveforms Gx(t), Gy(t) are preferable sufficient to cover their orbit uniformly to the k space. Not only a spiral, but also a zigzag orbit,

such as used in EPI method, can be used.

The RF waveform $B_1(t)$ calculated from these steps is saved to memory. And, the process is finished. Subsequently, MRSI is performed by using an RF excitation pulse modulated with this RF waveform $B_1(t)$ (step 15). In this step, the amplitude of the RF waveform $B_1(t)$ is arranged experimentally such that the maximum flip angle is 90°.

One embodiment of an MRSI imaging sequence according to the present invention will be described with reference to Fig. 3. In this imaging sequence, the spatial selective excitation is applied to the 3D-CSI method. At first excitation pulse 71 is applied together with gradient magnetic fields 74, 75 swinging, respectively, to the x and y directions. And, only the magnetization in the interior of the two-dimensional region of the x-y plane is excited. The slice selective gradient magnetic field Gz is not applied in this case. The interior of the ROI with nearly the shape of D(x,y) is excited with this excitation pulse. The accuracy of excitation depends on the amount of the high frequency component included in the gradient magnetic fields 74, 75 at the k-space. Subsequently, the spatial information of x, y is encoded relative to the phase of the magnetization by phase encoding gradient magnetic fields 76, 77. Refocus pulse 72 is applied together with slice selective gradient magnetic field Gz 73 to generate echo 78 after the TE interval from the start of excitation. Thus, the generated echo 78 is received as a signal.

And, waiting for the recovery of longitudinal

magnetization, the next cycle is performed to repeat signal measurement while changing the phase encoding amount. Three-dimensional Fourier transformation is performed on the acquired echo signal as a function of phase encoding (kx, ky) and the time t to obtain a spectroscopic image (metabolic distribution image). In addition, it is desirable for this MRSI method to use a spin echo type, as shown in the figure. Generally, when the frequency of the excitation pulse is offset from the resonance frequency, the shape of an excited region and the phase of the transverse magnetization is affected with this offset. But, in case of application of the spin echo type, the influence existed by this offset is canceled by application of refocus 180° pulse 72.

The reconstructed image of metabolic distribution can be displayed on a CRT. But, it is preferable to correct the spatial distribution by multiplying it with an inverse function of the excitation form function D(x,y). Then, a spatial variation of the flip angle in excitation is corrected to obtain a metabolic distribution image having fixed quantities.

As thus described, in the MRSI method of the present invention, there is no need for the OVS process for fat suppression. So, the measurement time of 3D-CSI having a double phase encode loop is shortened the same as having no suppression process. And also, fat suppression can be effectively achieved.

Fig. 5 is a view showing another embodiment of the 3D-CSI

method of the present invention. This imaging sequence is different from that shown in Fig.3 in that water suppression is used. That is to say, a water spectrum is excited by a known CHESS pulse 72 at each cycle of phase encoding, and it is dephased with application of a crusher gradient magnetic field 82. For CHESS pulse 81, a Sinc function or Gaussian function is used. They are appropriately selected in consideration of the form of the excited spectrum and the application time or the like.

The measurement following the application of the crusher gradient magnetic field 82 is the same as the imaging sequence shown in Fig.3. That is to say, the modulated waveform of the RF pulse is calculated based on the ROI that has been interactively preset on the scout image and saved to memory. And, this RF pulse modulated by the modulated wave form is applied together with the swinging gradient magnetic field in the directions x and y, respectively. And, the previously set ROI is excited selectively. And, after the phase encoding, a refocus pulse is applied to generate echo 78.

The previously described imaging sequence (two-dimensional MRSI) illustrated in Fig. 3 and Fig. 5 can produce a three-dimensional spectroscopic image by carrying out measurement while changing the slice position. In this case, when the ROI deviates corresponding to the slice position, the ROI can be moved in parallel from the reference position by applying phase modulation to the RF waveform $B_1(t)$.

That is to say, when the position of the ROI is shifted

about x0, the phase modulation, such as expressed in an equation (3), can be applied to the RF waveform.

$$B'_1(t)=B_1(t)e^{-ix0\cdot k(t)}$$

$$K(t)= -\gamma_t{}^T G(s)ds \qquad \ldots(3)$$

In equation (3), G(s) is a gradient magnetic field vector, and T is the application time of $B_1(t)$. With a specified shift x0 of the position on a scout image, the phase modulated waveform of the RF pulse can be calculated with the equation (3). In this case, the waveform of gradient magnetic field G(s) is predetermined.

Next, another embodiment of the MRSI method according to the present invention will be described with reference to the flow chart in Fig.6. Even in this embodiment, the method is the same as in Fig. 1 from measurement of the scout image (step 11) to calculation of the RF waveform $B_1(t)$ (step 14). But, in this embodiment, following the setting of the ROI on the displayed scout image, a setting of the image matrix size is performed (step 16,17). Thus, at first, after the ROI 61 is specified interactively on the image shown in Fig. 2(b), the smallest rectangular region 62 is automatically generated to cover the ROI (step 16). And, the image matrix size, for example 16 × 12, is set in accordance with this rectangular region (step 17). Imaging (spatial selective excitation MRSI) is performed with the phase encoding step (kx,ky) corresponding to this image matrix size.

In the normal 3D-CSI method, the phase encode step is usually performed with 16×16 or 32×32 size. As shown at 63 of

Fig. 2(b), its field of view is square. But, in this embodiment, the image matrix is set corresponding to the smallest rectangular region 62 covering the ROI. Accordingly, there is no need to measure exterior pixels where the magnetization is not excited, whereby the repetition time of the phase loop can be reduced. In addition, as shown in Fig. 2(c) and Fig. 2(d), it is preferable to rotate the direction of the coordinate axis of the image measuring matrix in accordance with the inclination of the ROI 61, that is, the direction of rectangle, as seen in Fig. 2(d). Thus, the region along the shape of ROI can be set, so it is possible to get the minimum number of repetition steps. The other features are the same as the embodiment shown in Fig.1.

In the above-described embodiment, an example of application of the invention to a two-dimensional MRSI has been described. But, it is possible to apply the invention to three-dimensional MRSI in the presence of a phase encoded gradient magnetic field of a selected slice direction. One example of a three-dimensional MRSI imaging sequence will be described with reference to Fig. 7. This imaging sequence includes water suppression (the application of CHESS pulse 81 and crusher gradient magnetic field 82) the same as the sequence shown in Fig. 5, and, moreover, a phase encode loop 91 in the slice direction is added to it. In addition, the gradient magnetic field in the slice direction, that is applied at the same time with refocus pulse 72, is a slab selective gradient magnetic field, and it can be omitted.

In this three-dimensional MRSI method also, the scout image is imaged in preprocessing, and the ROI is set on the scout image. And, the RF waveform is calculated from this shape function in the same way. And the image matrix size can be determined in the x and y directions in accordance with the set ROI.

In addition, in the above-mentioned embodiment, a spin echo type MRSI method has been described. And, it is indicated that the spin echo type is preferable for removing the influence of offset to the resonance frequency. But, in case of measuring material with a short T2, it is possible to employ an FID type.

This example is illustrated in Fig. 8. In Fig. 8, after the application of the RF excitation pulse 71, the phase encoded magnetic fields 76, 77 are applied in a short time. Then, the FID signal is measured immediately.

As thus described, the MRSI method with spatial selective excitation has been described by using the 3D-CSI method as an example. And, spatial selective excitation can be performed with a MRSI sequence by using another sequence, such as EPI, FSE or the like.

According to the present invention as described above, a necessary part of an object can be excited to obtain a signal while avoiding an unnecessary part thereof, such as fat. Then, a metabolic image can be obtained for precisely excluding fat signals. In addition, as a result of no need to suppress fat in an OVS of the preprocessing at each phase encoding, the

preprocessing can be shortened. And, a recovery of the water due to longitudinal relaxation can be reduced. In addition, desired ROI is set interactively on a scout image so as to automatically calculate the RF excitation waveform from the ROI shape with a calculator. And, MRSI is performed soon after the ROI is set. Furthermore, a minimum rectangular matrix is set in accordance with the ROI to reduce a useless measuring loop of the MRSI. As a result, speedup of the MRSI can be achieved.

These preferred embodiments of the present invention have been described with reference to the drawings. It will be easily understood that the present invention can be varied without departing from the scope of the invention.